

## New concepts and hypotheses

# Transfer of lumbosacral load to iliac bones and legs

## Part 1: Biomechanics of self-bracing of the sacroiliac joints and its significance for treatment and exercise

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### Summary

This study deals primarily with the stability of the base of the spine. The sacroiliac joints are vulnerable to shear loading on account of their predominantly flat surfaces. This raises the question of what mechanisms are brought into action to prevent dislocation of the sacroiliac joints when they are loaded by the weight of the upper part of the body and by trunk muscle forces. First a model is introduced to compare load transfer in joints with spherical and with flat joint surfaces. Next we consider a biomechanical model for the equilibrium of the sacrum under load, describing a self-bracing effect that protects the sacroiliac joints against shear according to 'the sacroiliac joint compression theory', which has been demonstrated *in vitro*. The model shows joint stability by the application of bending moments and the configuration of the pelvic arch. The model includes a large number of muscles (e.g. the gluteus maximus and piriformis muscles), ligaments (e.g. the sacrotuberous, sacrospinal, and dorsal and interosseous sacroiliac ligaments) as well as the coarse texture and the ridges and grooves of the joint surfaces.

### Relevance

In case of pelvic instability and peripartum pain several therapeutic approaches are used. Some of these approaches have a disabling effect themselves. At least in part, this is due to a lack of insight into the control of load transfer in the integrated system formed by the lower lumbar region, pelvis, and upper legs. The present study offers a theory on forces acting on the sacroiliac joints and the way their mobility is kept within small limits, which may contribute to a better understanding of the significance of muscle training and the use of a pelvic belt.

Key words: Biomechanics, sacroiliac joint, peripartum pain, sacrotuberous ligament, gluteus maximus muscle, symphysis.

### Introduction

It has been assumed that during pregnancy women bend the upper part of the body backwards to obtain equilibrium with the increasing weight in the abdomen. This would result in hollowing of the lumbar curvature,

which is supposed to be an important cause of frequent low back pain. With the help of an accurate measurement apparatus, however, it was demonstrated that after partus the curvature of the spine<sup>1</sup> in general becomes straighter<sup>1,2</sup> instead. These measurements affirmed earlier observations of Kuhn<sup>3</sup> and Swanson<sup>4</sup>, who measured lordotic depth, and of Weltz<sup>5</sup>, who compared pictures. Another possibility is that the cause of pain during or after pregnancy is related to mechanical problems in the pelvis and the joints related directly. Therefore research was started on the functional anatomy of the sacroiliac joints. This research included biomechanical modelling,

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verification by experiments on human cadavers and mechanical models, tests on healthy subjects, and patient studies. In this article, our attention will be focused on biomechanical modelling.

The angular movements in the sacroiliac (SI) joints are small compared with the joint excursions in the lumbar vertebral segments. Therefore the biomechanical focus was directed towards the solidity of the structures that form the basis of the spine. It was demonstrated that the SI joints normally have a macroscopic appearance that allows small displacements even at old age<sup>6</sup>. These findings agree with other studies showing that displacements in the SI joints are present<sup>7-11</sup> even at old age<sup>6</sup>.

A common clinical observation is the increased laxity of ligaments in the pubic area during the later stages of pregnancy and the postpartum period<sup>12</sup>, which may be relevant for patients suffering from peripartum pain. This pain can be severe, the onset can occur during normal activities like sitting and walking and it may result in chronic illness and disability. The biomechanical model introduced in this study is aimed at a better understanding of how women, depending on their impairment, can be sufficiently trained to recover.

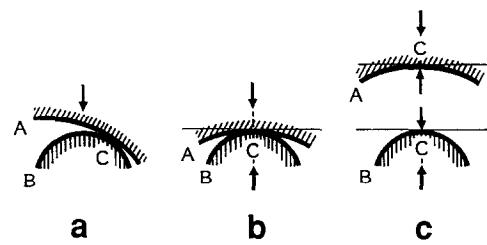
The literature on biomechanics of the sacroiliac joints generally concerns kinematics, i.e. the determination of axes of rotation and geometry<sup>7,9,13</sup>. These studies do not present biomechanical models on SI joint stability, with special reference to the typical properties of the SI joints. The model in the present article indicates that the SI joint can sustain large bending moments and compression, but is vulnerable to shear. This leads to the hypothesis that subluxation due to large shear loading is counteracted by compression of the SI joints.

The literature on the biomechanics of load transfer across the pelvis includes the concept of the 'pelvic ring'<sup>14-16</sup> and of the 'pelvic arch'<sup>17</sup>. Both concepts will have significance in different loading modes. However, the present study supports the hypothesis that the transfer of large lumbosacral loads to iliac bones and legs relies on the presence of a stable pelvic arch.

### Shape of the sacroiliac joint surfaces

The loading mode of joints can be transverse, tensile, compressive, bending or torsional. Spherical and flat joint surfaces have different capabilities in transferring these respective loading modes. In tension and torsion the form of the joint surfaces has no influence, as long as they are purely spherical or purely flat. In compression, the best conformity of adjacent surfaces gives the best capability of load transfer. The type of joint surface becomes of interest when comparing transverse forces and bending.

In Figure 1a the joint surfaces make contact in the cartilage contact point C. It can be demonstrated that the force drawn in the picture leads to rolling and gliding before equilibrium is reached (Figure 1b). The joint reaction force must run through the contact point

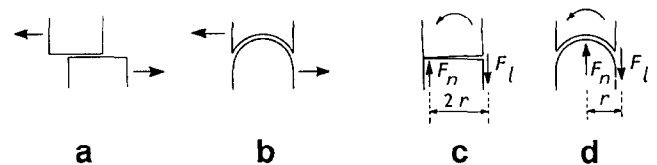


**Figure 1.** a Situation of rolling and gliding of bone A with respect to bone B. b, c Direction of the joint reaction force is perpendicular to the tangent to the joint surfaces at the point of cartilage contact.

(which represents a contact area). Further, when friction can be neglected, the joint reaction force must be perpendicular to the common tangent plane to the joint surfaces. In Figure 1c the surfaces are separated, illustrating the line of action of the joint reaction force being perpendicular to the joint surface.

The previous principle is applied in Figure 2, showing a joint with flat and a joint with spherical (or cylindrical) surfaces. Application of a transverse force near the joint in Figure 2a leads to a shift of the upper bone with respect to its adjacent bone until this shear is stopped by ligaments and/or muscles. The bones do not remain in line, which points to the risk of (sub)luxation. This significant disadvantage does not exist in spherical joints (Figure 2b). In Figure 2c the flat joint is loaded with a pure bending moment. The joint cavity tends to a wedge shape and the bone contact point shifts to the edge. The bone contact force ( $F_n$ ) is perpendicular to the joint surface and equal to the ligament force ( $F_l$ ). In Figure 2d the bone contact force, which must be parallel to the force from ligaments and/or muscles ( $F_l$ ), cannot shift to the edge and remains on top of the sphere. The lever arm of the couple formed by  $F_n$  and  $F_l$  in the joint with the flat surface (Figure 2c) is about twice as large as that in the ball-and-socket joint (Figure 2d). We can conclude that a joint with predominantly flat surfaces is well suited to transfer great moments of force but is vulnerable to transverse forces near the joint. Therefore, flat surfaces go with restricted joint excursions (see also Appendix A).

The foregoing principles may well be appreciated to the predominantly flat joint surfaces of the SI joints. The contour of the sacral surface of the SI joint shown



**Figure 2.** a Subjecting a bone to a transverse force near the joint leads to a shift with respect to the adjacent bone before it is stopped by ligaments. The bones do not 'stay in line', which points to vulnerability to (sub)luxation by this loading mode. b This advantage does not occur with a ball-and-socket joint. c, d Because of a greater lever arm a flat joint is more appropriate than a ball-and-socket joint to transfer a pure bending moment.

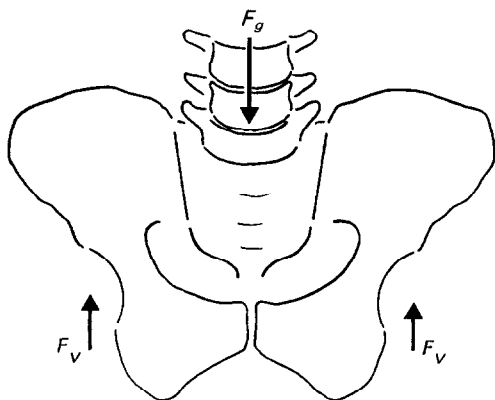
in Appendix B (Figure B1) illustrates that the dimension ( $h$ ) in the longitudinal direction of the sacrum is approximately twice as large as the dimension ( $b$ ) in the transversal direction (see Table B1). The advantage of this shape (in combination with adjacent ligaments that cross the joint) compared to a circular shape with a similar area, concerns a favourable resistance against bending moments in planes that practically coincide with the largest dimension ( $h$ ). This dimension has also significance for the following concept of the 'pelvic arch'.

### Self-bracing mechanism against shear

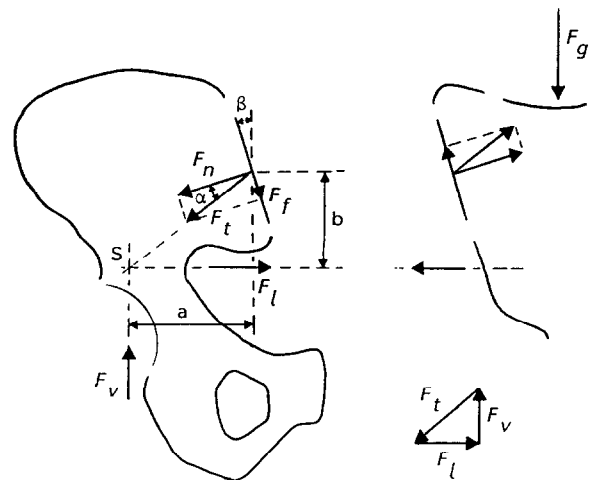
The flat surfaces of the SI joint in combination with its ligamentous system (a.o. interosseous and dorsal SI joint ligaments) point to a favourable intrinsic joint stability for the transfer of bending moments. However, the flat surfaces make the SI joint vulnerable to dislocation by shear. The loading mode in Figure 3 raises the question of what mechanism prevents the caudal shift of the sacrum in relation to the iliac bones. A hypothesis is formulated that subluxation due to large shear loading is counteracted by compression of the SI joints. This compression is caused by a self-bracing mechanism which comprises (a) the loading mode of an arch, and (b) muscle and ligament forces practically perpendicular to the joint surfaces. In short, the foregoing can be defined as 'the sacroiliac joint compression theory'. The model starts from a person in an upright position with equal loads on both legs (Figure 3). The weight of the suprasacral part of the body, about 60% of the total body weight, is carried by the sacrum. The vertical force vector  $F_g$  coincides with the longitudinal axis. The weight of the sacrum is not taken into account.

#### The loading mode of an arch

The undulated shape of the articular surfaces of the SI joint and the spatial configuration of adjacent bones are reduced to the model of Figure 4, which deals with equilibrium of the sacrum and the ilium in an upright



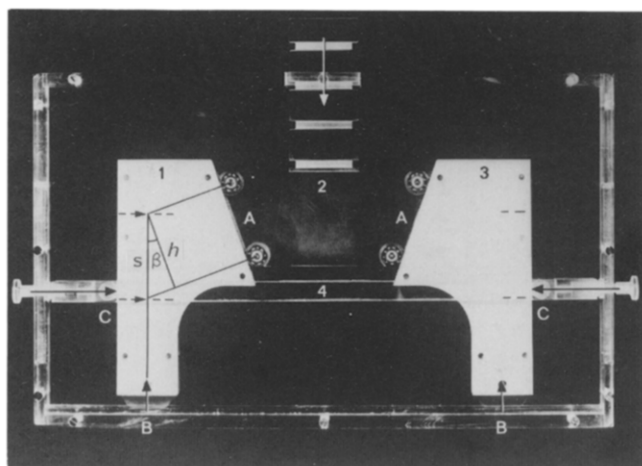
**Figure 3.** Simplified loading mode of the pelvis due to gravity. Each leg carries half of the weight ( $F_v$ ) of the suprasacral part of the body ( $F_g$ ).



**Figure 4.** Model of self-bracing effect, stabilizing the SI joints in the plane of drawing. The lines of action of three forces intersect at one point (S). Force  $F_t$  may be raised by ligaments, muscles, or a pelvic belt just cranial to the greater trochanter and caudal to the SI joint. Self-bracing is reinforced by enlarging the friction force  $F_f$  by means of enlarging the normal force  $F_n$  with the help of muscle tension. The gluteus maximus muscle has the proper direction.

position with equal loads on both legs. The joint reaction force  $F_t$  can be resolved into a force perpendicular to the joint surface ( $F_n$ ) and a tangential force ( $F_f$ ), the latter resulting from friction. The coefficient of friction equals  $\tan \alpha$ , which is the ratio  $F_f/F_n$  and refers to the friction force just before sliding occurs. The wedge shape of the sacrum is represented by the wedge angle  $\beta$ . The lines of action of the joint reaction force ( $F_t$ ) and the hip joint force ( $F_v$ ) intersect in point S. Equilibrium will be reached independent of lumbosacral load by the configuration of an arch, which involves a third force ( $F_l$ ) that also passes through S and represents ligament or muscle forces. The triangle shows the mutual proportions of the three forces. The position of point S changes with friction (angle  $\alpha$ ), with a point of application of the SI joint reaction force at a more cranial or a more caudal site, and with other levels and orientations of ligaments and muscles.

With the aim of demonstrating the self-bracing phenomenon, a mechanical model has been made. In Figure 5 three plastic parts form a composition resembling sacrum and iliac bones. Friction between the contact surfaces has been made negligible by means of ball bearings at A and wheels at B. This forms a loose construction which can be made rigid by the adjustment of a tensile strip (4). This strip can be translated up and down to demonstrate that this mechanism only exists within geometrical boundaries. These boundaries are determined by the size of the joint, which is illustrated by the large dimension  $h$  in the Appendix (Figure B1). In Figure 5 this results in a range with the height  $s = h/\cos \beta$ . When the strip is translated to a level which is too high or too low (see dotted lines), the construction will collapse. By adding friction, this range ( $s$ ) becomes larger. This will happen



**Figure 5.** Composition of three parts (1, 2, 3) held together by a tensile strip (4). This strip can be freely translated up and down to demonstrate that at a level too low or too high the construction will collapse. These levels are indicated by arrows on a distance  $s = h/\cos \beta$ . At the contact surfaces (A) friction is eliminated by means of ball bearings and at B by means of wheels. The configuration remains stable also after the addition of extra lumbosacral load.

when the ball bearings at A (Figure 5) are removed (see Appendix D).

In Appendix C, the configuration of the sacrum and ilium is roughly compared with the arch of the foot, which provides for compressive loading of the midtarsal joint in Figure C3 and avoids shear. In the model of Figure 4 the symphysis is absent. This indicates that the concept of the 'pelvic ring' is not required for the modelling of the transfer of large lumbosacral loads to the legs. Evidence may be found in the X-rays (Figure 6) of a woman, age 34, before and after resection of the pubic bone because of a chondrosarcoma. In a follow-up period of 4 years the mechanical function remained normal without complaints.

#### *Stability by ligaments, muscle forces, and pelvic belt*

The force  $F_l$  in Figure 4 may be produced by ligaments, e.g. by a force component from the sacrotuberous ligament, and possibly the (deep) fascia in the region. However, we can also think of tension in the gluteus maximus, piriformis and coccygeus muscles. Since we noted positive results of the application of a pelvic belt in women with peripartum pain, we propose the hypothesis that the belt force acts like  $F_l$ .

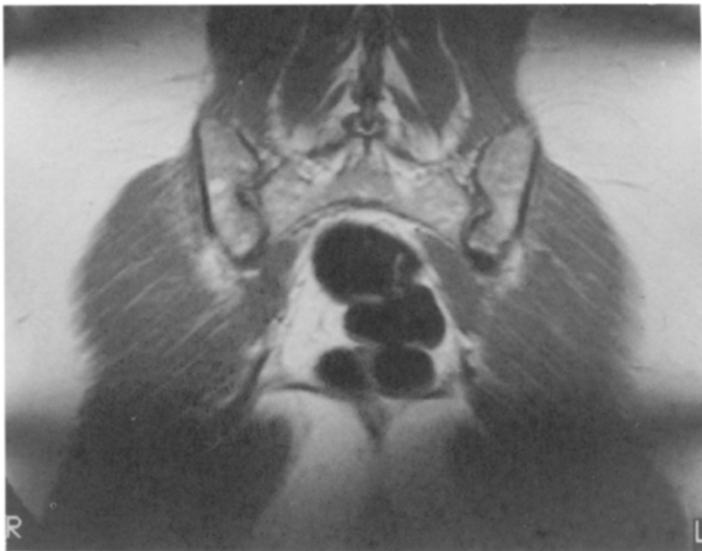
A further observation is that the self-bracing mechanism in Figure 4 is facilitated by enlarging the friction between the joint surfaces, which can be achieved by a coarse texture<sup>18,19</sup>. On eight embalmed SI joint surfaces without ridges and grooves a friction coefficient of 0.4 (SD 0.2) was measured. Resistance against sliding can also be increased by ridges and grooves in the joint surfaces, which may be interpreted as an increase of the wedge angle  $\beta$ . The effect of ridges and grooves and of friction can be increased by enlarging the normal force  $F_n$ . Here we think of a

tension in muscles with a force component practically perpendicular to the SI joint, in particular the gluteus maximus and piriformis muscles. The magnetic resonance imaging (MRI) picture of Figure 7 shows the orientation of gluteus maximus fibres in relation to the SI joints. Tension in the gluteus muscles can result in SI joint compression because these muscles insert, in part, into layers of the external fascia superficial to the sacrum with connections to the thoracolumbar fascia.

In a sitting posture, the hip joint forces ( $F_v$  in Figure 4) are replaced by forces acting on the ischial tuberosities. This means that the lever arm  $a$ , which is an essential part of the arch mechanism, becomes smaller or even zero. In prolonged sitting with a backrest (i.e. passive sitting) not only is the arch absent, but also the self-bracing muscle activity as mentioned above is missing.



**Figure 6.** Radiographs of a woman, age 34, **a** before, and **b** after resection (see arrows) of the pubic bone because of a chondrosarcoma. Mechanical function remained normal. (Courtesy of B. van Linge MD, PhD.)



**Figure 7.** MRI picture of a female sacroiliac joint, age 42. This coronal slice clearly shows the orientation of gluteus maximus muscle fibres in relation to the SI joints.

When self-bracing against shear according to pelvic arch and activity of muscles is absent, the feature of the transfer of large bending moments by flat joints may be especially relevant. A bending moment goes with joint compression (see Figure 2c), which in combination with a coarse texture and ridges and grooves results in resistance against sliding. An example of this could be sitting on a firm seat on one ischial tuberosity with the ipsilateral upper leg crossed upon the other leg. The moment of force about the SI joint above the ischial tuberosity results from the weight of the upper part of the body and leg. Here, tension in the large dorsocranial SI ligaments can be expected. When sitting on both ischial tuberosities, a similar stabilizing moment of force on the SI joints can be produced by placing a foot on the ipsilateral upper leg. By the weight of the leg the hip joint is pushed into abduction.

Crossing the legs may also contribute to SI joint compression by stretching dorsal ligaments and thoracolumbar fascia as a result of hip flexion and hip adduction. Although many reasons may exist to cross the leg in sitting, no literature is known to us giving a biomechanical explanation.

#### *Muscles that enhance SI joint load and muscles that contribute to self-bracing*

In Figure 4 the self-bracing effect comprises the following force-parameters:

- (a) the load imposed on the sacrum ( $F_g$ ),
- (b) the friction between joint surfaces obtained by a normal force ( $F_n$ ) and
- (c) the ligament or muscle force ( $F_l$ ) to obtain a self-bracing effect within certain geometrical boundaries.

A further description of the structures involved can be useful. This leads to the following additional

observations on  $F_g$ ,  $F_n$  and  $F_l$  respectively:

$F_g$ : This force is produced by the weight of the upper part of the body. However, the load on the sacrum will be considerably enlarged by the action of several muscles. These actions depend on body posture, which will be dealt with in part 2 of this article. The following muscles can be of influence: the erector spinae, quadratus lumborum, obliquus abdominus, rectus abdominus, latissimus dorsi, and psoas major.

$F_n$ : The contributions of the gluteus maximus and piriformis muscle must be mentioned as the most important ones. Furthermore, minor force components of some inner pelvic and iliolumbar muscles can be expected.

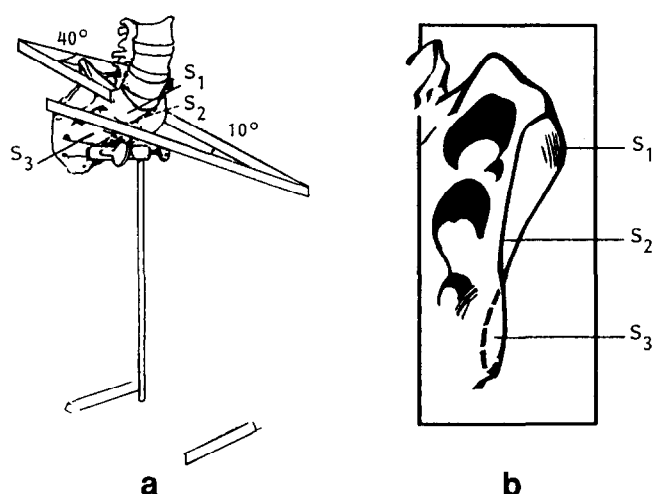
$F_l$ : Here a contribution can be expected from the sacrotuberous ligament, the tension of which can be influenced by the biceps femoris<sup>20,21</sup>, and also from the m. piriformis, the m. coccygeus, and the caudal track of the m. gluteus maximus.

#### *SI joint stability in the sagittal plane*

The hypothesis concerning the essential role of compression for the stability of the sacroiliac joints has been applied to a loading situation in an anteroposterior view of the pelvis. The same hypothesis, including the model of the pelvic arch, has significance for joint stability in the sagittal plane, where flexion and extension of the sacrum are of interest. Such rotations involve sliding of joint surfaces which are counteracted by compression, enhancing friction and the resistance of ridges and grooves. Furthermore, the different wedge angles in transverse cross-sections at the cranial and caudal side of the SI joint point to a slight twist resembling the shape of a propeller blade<sup>30</sup> (Figure 8). As such, two SI joint surfaces can counteract flexion in relation to the hip bones, provided that they are pressed together. This, again, appeals to compression and fits in the model on self-bracing. Because in transversal caudal cross-sections the wedge angle of the sacrum is open in ventral direction (see part 2, Figure 5b), this form will not counteract extension of the sacrum in relation to the hip bones. So it can be assumed that the potential resistance to extension is less than the potential resistance to flexion.

#### **Discussion**

The SI joint can be compared with a multidirectional force transducer, for it can be expected that the ligaments surrounding the SI joints possess mechanoreceptors. In that case, the SI joints can be assumed to be large sensors located in the middle of considerable force streams being transferred by the pelvis from the upper part of the body to the legs. In this transfer the largest muscles of the body are involved. In terms of sensitivity for local stresses only small SI joint excursions are required, as is the case with strain gauges<sup>22,23</sup>. On cadavers it was



**Figure 8.** a, Ventrolateral view of sacrum showing the angle between the sacral auricular surfaces of the SI joint at the level of S<sub>1</sub> and S<sub>3</sub>. b, Schematic drawing of the sacral auricular surface of the right SI joint showing the generalized orientation, which resembles the shape of a propeller. (Reprinted with permission from Dijkstra et al. 1989<sup>30</sup>.)

demonstrated<sup>6</sup> that such small excursions occur up to old age. Since our research focuses on the control of load transfer in the integrated system formed by the lower lumbar region, pelvis, and upper legs, we are especially interested in the forces acting on the SI joints and the way their mobility is kept within small limits. So the aim of this study was to find anatomical and biomechanical principles that contribute to the control of the stability of the SI joints.

Because of the predominantly flat shape and the conformity of adjacent sacral and iliac joint surfaces, the SI joint is well designed for the transfer of great moments of force and compression. In comparison with other flat joints like the tarsal and carpal joints, only restricted mobility may be expected. The great advantages of a flat joint, as mentioned before, go together with vulnerability to (sub)luxation due to shear. With a view to a defence mechanism against shear, we introduced hypotheses on biomechanical principles which lead to (a) a loading mode of compression and bending that avoids considerable shear, and (b) resistance against sliding of the joint surfaces.

(a) Starting with the similarity in flat shape of the SI joint and the midtarsal joint, we suggest an analogy to load transfer in the arch of the foot. The construction of an arch provides for a loading mode of joint compression. In the foot an essential role is attributed to the tension in the plantar aponeurosis, which is raised by dorsiflexion of toes and leads to a close-packed position of the tarsal bones at heel strike and during push off<sup>24</sup>. In the pelvis we observe a similar mechanism which ascribes a role to the sacrotuberous ligament, where tension can be increased by the biceps femoris muscle<sup>20,21</sup>, and to the piriformis, the coccygeus, and the gluteus maximus muscle. With the use of geometrical and force parameters based on

anatomical studies (Figure 4) we analysed this mechanical principle and called it self-bracing of the SI joint. With the model of Figure 4 SI joint stability can be demonstrated without a contribution from the symphysis. We point out that a small deformation of the symphysis is essential to allow a minute but free adjustability of the SI joint surfaces for load transfer, like the slight tendency to wedge formation corresponding to Figure 2c. In this respect isolated symphysiodesis may be questioned<sup>25</sup>.

(b) Resistance to sliding of flat surfaces can be obtained by, (b1) roughening of the contact area which increases the friction coefficient, (b2) development of grooves and ridges, and (b3) a loading mode of high compression.

(b1) The texture of the joint surface is not smooth. *In vitro* we measured friction coefficients in the order of magnitude of 0.4<sup>18,19</sup>, which is considerable.

(b2) It is obvious that grooves and ridges form a resistance to sliding. A ridge, however, can also be assumed to be a contribution to the (local) increase of the wedge angle of the sacrum.

(b3) An important provision against sliding may be the generation of compression by muscles crossing the joint. Here an important role can be ascribed to the gluteus maximus muscle. With regard to the lumbodorsal fascia we found that caudal fibres of the fascia can show continuation to the heterolateral side. This could imply that forces generated, e.g. in the latissimus dorsi, could function synergistically with the heterolateral gluteus maximus muscle in certain loading conditions.

The different mechanisms that warrant stability of the SI joints may become less effective due to the decline of muscle performance and/or increased laxity of ligaments. Decline of muscle function can occur in people withdrawing from sports, undertaking sedentary work, etc. Increase of laxity especially occurs peripartum<sup>25</sup>. With respect to patients suffering from peripartum pain, the assumption is made that hypermobility in the SI joints can cause serious impairments. A method to restore pelvic stability is the use of a pelvic belt<sup>1</sup>. The fact that using such a belt can reduce pain may be the reason why in several cultures (e.g. Indonesia, Turkey, Morocco) an elastic corset is worn at about the S<sub>2</sub>-level from the 6th month of pregnancy onwards. According to our biomechanical model we conclude that if such a belt is put on with a small force resembling the force in laces to close a shoe<sup>26</sup>, this will be sufficient to generate a self-bracing effect in the SI joints under heavy load. A large belt force is not recommended, because it can cause irritation and oedema and it may be detrimental to the symphysis. In a loading experiment on human preparations, flexion of the sacrum in relation to the hip bones decreased by about 20% when applying a belt force of only 50 N<sup>25</sup>. A wide and pliable (but inextensible) belt which does not irritate the thighs in sitting posture is advised. Because the model points to a

modest belt force, the additional metal plates which were introduced on the market are rejected. The biomechanical model also indicates that the belt must be positioned just cranial to the greater trochanter and caudal to the SI joints. A position of the belt caudal to the SI joints has the added advantage that it can counteract flexion by pressure on the caudal–dorsal side of the sacrum. Another approach to counteract hypermobility of the SI joints is the training of muscles.

The study on the stability of the SI joints leads us to a discussion about symphysiolysis. It was suggested that symphysiolysis is connected to hypermobility of the SI joints<sup>25</sup>. Consequently, isolated surgical symphysiodesis can be questioned because it does not resolve the problem of SI joint instability.

Further research will be necessary to quantify the parameters that play a role in the biomechanical principles presented in this study and to translate the findings of the biomechanical modelling into validated exercise protocols. There is also a need for a consistent definition of pelvic hypermobility and an objective method for its diagnosis. As to this, the schematic and separate presentation of biomechanical principles that participate in the self-bracing of the SI joints may form a basic contribution.

## Conclusions

In this paper the typical properties of the SI joint are of interest. In this respect we can summarize the following conclusions:

- Because of the predominantly flat shape the SI joint is suited for the transfer of large bending moments and compression. However, flat joints are in principle vulnerable to shear forces.
- The difference between the length of the SI joint in the longitudinal direction of the sacrum and that in the direction perpendicular to it indicates that this joint is strongest in a longitudinal direction. Regarding this we estimated a theoretical efficiency factor.
- Flat joints can be loaded with compression while shear can be avoided. To elucidate this, we introduced an analogy between the sacroiliac and the tarsal joints, as regards the loading mode of an arch.
- A free body diagram of the sacrum and of the iliac bones, being parts of the pelvic arch, resulted in a description of 'a self-bracing mechanism'. With the help of a mechanical model we were able to demonstrate when this system is stable and when it will collapse.
- Parameters in the model on self-bracing are friction, wedge angle, joint compression, and the geometry of an arch held by ligaments and muscles. Friction and wedge angle can be enlarged by local ridges and grooves and a coarse texture of joint surfaces.
- The model of the pelvic arch (which supports the hypothesis on the essential role of compression to provide for joint stability) also has significance for joint stability in the sagittal plane. In this plane, flexion and extension in relation to the hip bones are of interest.

With reference to the slightly twisted form of the joint surfaces, resembling the shape of a propeller, less potential stability is ascribed to extension.

- We marked muscles and ligaments that contribute to self-bracing. An important role can be ascribed to the gluteus maximus muscle and to the sacrotuberous ligament. With anatomical preparations it was demonstrated that tension in the sacrotuberous ligaments can be enhanced by pull from the biceps femoris and gluteus maximus muscles and parts of the deep layer of the lumbodorsal fascia.

- An explanation of the positive experiences with the application of a pelvic belt to relieve peripartum pain can be that its action fits in the model on self-bracing of the SI joints. So the belt can be put on with small force, resembling the action of the laces of a shoe; its position should be just cranial to the greater trochanter and caudal to the SI joints. The line of action of the piriformis muscle is compared to that of the pelvic belt.

- In passive sitting with a back rest on a firm seat, the SI joint stabilization by (most of) the muscle forces and the arc are absent. This represents one of the situations where a bending moment of force about the SI joints could produce joint compression against shear. We speculate that sitting with crossed legs can provide for the necessary moment of force.

- In the model of self-bracing the symphysis does not play a role. It is assumed, that the symphysis must be slightly elastic as to permit minute but free wedge formation of the SI joint cavity in all directions for the adjustment to alternating load. In this respect, isolated surgical symphysiodesis can be questioned.

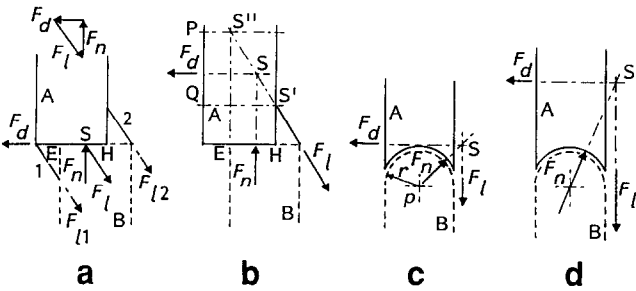
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## Appendix A: Spherical and flat joint surfaces

The different joints in the human body have a wide variety of often highly irregular forms. To elucidate relations between form and function, two extreme forms of joint will be discussed: the plane joint and the ball-and-socket joint. In Figure A1a bone A is loaded with a transverse force  $F_d$  at the level of the joint. At first bone A must shift with respect to bone B to produce the necessary stress in the ligaments which stops the shear movement. Bone A will not tilt when ligaments 1 and 2 produce a resultant force  $F_l$  which intersects the joint surface between the points E and H. Now the resultant joint reaction force  $F_n$  can intersect point S as well, while it acts perpendicular to the joint surfaces when friction is neglected. The proportion between the forces  $F_d$ ,  $F_l$ , and  $F_n$  is found by construction of the closed triangle of forces. In Figure A1b, the transverse force  $F_d$  acts on bone A higher up, between P and Q. Stressing one ligament now is sufficient to ensure that bone A does not slide further



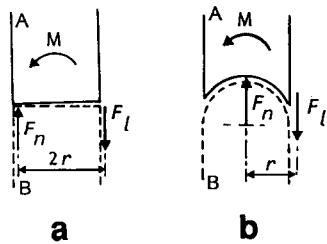


**Figure A1.** Transverse force acting on a flat and on a ball-and-socket joint.

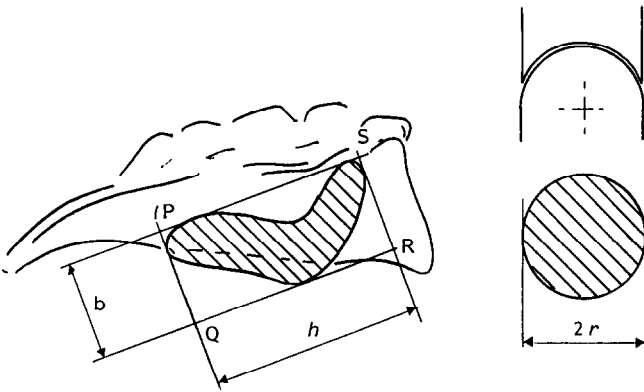
and tilt. As long as  $F_d$  lies between P and Q, equilibrium can be reached with all three forces acting on bone A passing through a point somewhere between  $S'$  and  $S''$ . If  $F_d$  acts at a higher level than P, equilibrium of bone A can only be obtained after tilting on edge E of bone B. In Figure A1c the transverse force is applied at the level of a ball-and-socket joint, which, after some rotation, gives rise to a ligament force  $F_l$ . The joint reaction force  $F_n$  passes through pole p, the centre of the circular cross-section of the joint surfaces, since the normal to any tangent to a circle always passes through its centre. The direction of  $F_n$  is given by the line through point S, which is the point of intersection of  $F_d$  and  $F_l$ . The magnitude of  $F_n$  can be found from a triangle of forces. In Figure A1d, force  $F_d$  is applied some distance from the joint. The point of intersection of S will then also be at this level. Hence  $F_n$  will point more in the direction of the longitudinal axis of the bone, and both  $F_l$  and  $F_n$  will be larger. In other words the two joint surfaces are pressed together harder when a given transverse force is applied further from the joint.

In Figure A2a the plane joint transfers a pure bending moment. After a slight tilt of bone A the joint reaction force shifts to the edge, which results in a large lever arm for the couple formed together with ligament force  $F_l$ . Only about half of this lever arm is formed in the case of the ball-and-socket joint of Figure A2b.

The above approach shows marked differences in the response of ball-and-socket and plane joint to transverse forces and bending moments. The plane joint is vulnerable to (sub)luxation while the spherical or cylindrical joint is less suited for the transfer of great moments of force. However, the latter disadvantage can be removed by enlarging the lever arm of ligament or muscle force  $F_l$  with respect to the joint axis by



**Figure A2.** Moment of force (M) acting on a flat and on a ball-and-socket joint.



**Figure B1.** a Tangents drawn to the contour of an SI joint give an impression of lever arms comparable with Figure A2a. b A ball-and-socket joint with an equal projected area (shaded) would have less capability for the transfer of moments of force.

means of a protuberance. An example of this is the trochanter major (see Figure C1).

### Appendix B: Shape of SI joint

In Figure B1a contours are drawn of the sacral side of the SI joint as identified on a macerated sacrum, without cartilage. We focused our attention to this material because of former difficulties in establishing exactly the border between fibrous apparatus and cartilage when we attempted to confirm the details of Bakland and Hansen's observations<sup>27</sup> on the geometry of auricular and axial areas.

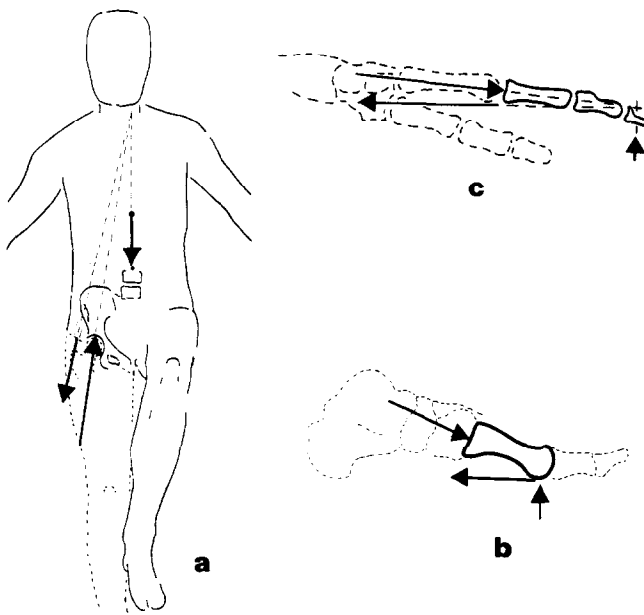
To the auricular contour, tangents are drawn resulting in a large lever arm ( $h$ ) and a smaller lever arm ( $b$ ). The meaning of these lever arms is similar to the width of the bones in Figure 2c. The different magnitudes indicate in which direction the largest moments of force can be transferred.

First it can be noticed that the auricular surface (shaded area) is smaller than the area formed by the quadrangle PQRS. By measuring the respective areas of eleven sacral auricular surfaces on six pelves we

**Table B1.** Dimensionless quotients determined on the sacral sides of eleven SI joint surfaces theoretically indicating 'efficiency factors'. Referring to Figure B1 the parameters  $h$  and  $b$  are the largest and the smallest width respectively. The projected areas of the ball-and-socket joint and of the SI joint are equal (shaded).  $A_{SI}$  is the projected area of the SI joint (shaded) and  $A_{PQRS}$  is the area of the respective quadrangle. Measurements refer to six pelves with an average maximal width at the iliac crests of 23 cm (SD = 0.8 cm).

	$\bar{x}$	SD
$\frac{A_{SI}}{A_{PQRS}}$	0.6	0.04
$\frac{h}{b}$	1.9	0.23
$\frac{h}{r}$	3.1	0.14



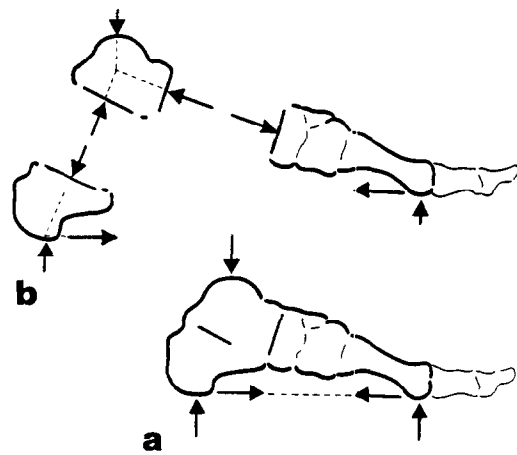


**Figure C1.** The arrangement of bones, muscles and joints leads to joint reaction forces in almost longitudinal direction of the respective bones. Equilibrium exists when three forces in the plane of drawing intersect at one point. The magnitudes of the forces can be constructed by means of a triangle (Appendix A).

found a value of circa 60% (Table B1), which may be interpreted as an efficiency factor. Another efficiency factor is formed when the flat auricular joint is compared with a ball-and-socket joint, both having the same projected area (Figure B1b). Referring to Figure 2c and d, for bending an efficiency factor 2 can be expected between the lever arms  $h$  and  $r$ . For the SI joints, however, we determined an efficiency factor  $h/r$  of approximately 3 (Table B1). Although the foregoing efficiency factors are dimensionless quotients, the recorded values still may be different in persons of different stature and race. The pelvis we measured had an average maximum width at the crista iliaca of only 23cm. The respective body heights were unknown.

#### Appendix C: Equilibrium of bones under load; compression of joints

In Figure C1 joint reaction forces are shown due to load from gravity (hip joint<sup>12</sup>, and metatarsal bone<sup>28</sup>), and from an external force on a finger<sup>29</sup>. In these three cases the mechanism is such that the joint reaction force can act in the direction of a line perpendicular to a tangent on the joint surface. This applies to a joint reaction force being oriented as much as possible in the longitudinal direction of the respective bone. In Figure C2a, a simplified model of the loading of the arch of the foot is given. In Figure C2b, the equilibrium of the talus, the calcaneus, and the combination of the other tarsal and metatarsal bones are given separately. This illustrates that the predominantly flat surfaces of the respective tarsal joints in this model transfer compressive forces as a result of tension in the plantar aponeurosis, the calcaneonavicular ligament, the



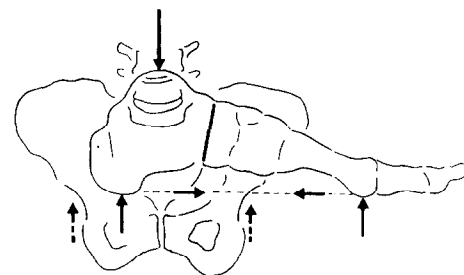
**Figure C2.** a A simplified model of load transfer by the arch of the foot. b Plantar tensile forces provide for compression of the tarsal joints. This principle, resembling a Roman arch, helps to avoid shear in the direction of the flat joint surfaces.

plantar ligaments, and the plantar muscles crossing the joints. Since joints with flat surfaces are vulnerable to transverse forces, the mechanism of Figure C2 is helpful to prevent shear-loading of such joints. This principle, resembling a Roman arch of stones resting on immovable piers, may be applicable to the self-bracing of the SI joints.

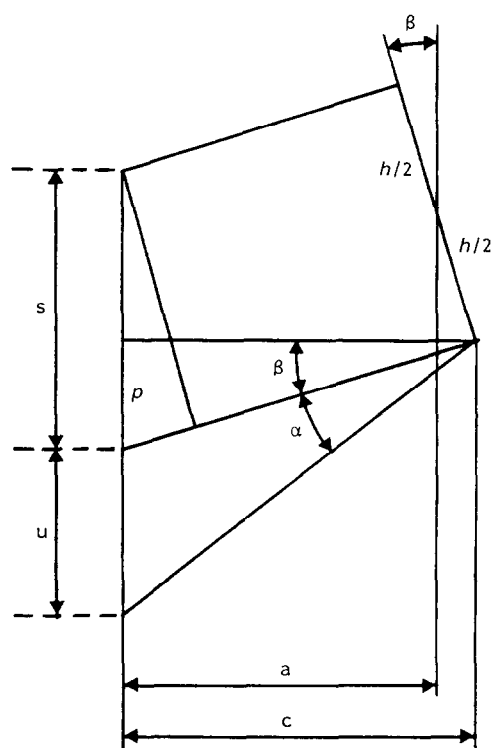
An analogy between a predominantly flat tarsal joint, for example the midtarsal joint, and the SI joint is illustrated in Figure C3. It shows that both joints are especially capable of transferring large compression forces and bending moments. In this analogy the pelvic arch has to be supported by muscles and ligaments as well. The hypothetical pull of a pelvic belt just cranial to the greater trochanter can also be recognized in the drawing.

#### Appendix D

In Figure D1 the distance  $a$  refers to Figure 4, and the measure  $h$  to Figures 5 and B1. In Figure 5 the range with stable positions  $s = h/\cos \beta$ . Adding a friction coefficient which equals  $\tan \alpha$  leads to increase of the



**Figure C3.** An analogy between the load on the SI joint and the load on a tarsal joint. Joints with flat surfaces are capable of transferring considerable compressive forces and bending moments. Tensile forces in ligaments and muscles can counteract shear in the direction of the joint surfaces (see Figure C2). The (horizontal) line of action of a pelvic belt just cranial to the greater trochanter can also be recognized.



**Figure D1.** The range ( $s$ ) with stable positions in Figure 5 is increased to ( $s + u$ ) by adding friction. With reference to Figure 4 the friction coefficient  $\tan \alpha = F_f/F_N$ .

range  $s$  to the range  $s + u$  (Figure D1). The expression for  $u$  can be derived as follows:

$$\begin{aligned} p &= c \tan \beta \\ p + u &= c \tan (\alpha + \beta) \\ u &= c(\tan (\alpha + \beta) - \tan \beta) \\ c - a &= h/2 \sin \beta \\ u &= (a + h/2 \sin \beta) (\tan (\alpha + \beta) - \tan \beta) \end{aligned}$$

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